RESPIRATION MONITOR FOR COMPUTED TOMOGRAPHY

The following relates to the diagnostic imaging arts. It finds particular application in respiratory compensated or gated diagnostic imaging, and will be described with particular reference thereto. However, it will also find application in conjunction with various diagnostic imaging modalities, such as computed tomography, single photon emission computed tomography,

5   positron emission tomography, magnetic resonance imaging, and the like.

In situ monitoring of patient respiration during medical diagnostic imaging advantageously facilitates image registration over the respiratory cycle, sorting or binning of acquired imaging data with respect to respiratory state, data filtering based on respiratory state, prospective respiratory gating, radiation dose modulation synchronized with the respiratory cycle,

10   and the like. Respiratory monitoring can also be useful for automating imaging. For example, initiation of imaging can be triggered by detection of the start of a patient breath-hold.

Heretofore, patient respiration has been monitored by electromechanical transducers that monitor chest movement. These electromechanical devices are relatively bulky, can feel uncomfortably restrictive to the patient, and may inhibit chest movement and respiration.

15   In one example, an air filled tube is wrapped around the patient's torso. Inhaling increases pressure in the tube, which is sensed by an associated pressure transducer.

Moreover, electromechanical respiration monitors can interfere with imaging, depending upon the transducer materials and the imaging modality employed. Fitting the patient with an electromechanical transducer is time-consuming and adds complexity to the preparatory

20   phase of the imaging session. In the case of cardiac or other types of imaging in which both the respiratory cycle and the cardiac cycle are monitored, preparation for imaging includes fitting the patient with both the electromechanical respiration monitor and electrodes of an electrocardiograph.

The present invention contemplates an improved apparatus and method that

25   overcomes the aforementioned limitations and others.

According to one aspect, a diagnostic imaging system is disclosed. A diagnostic imaging scanner acquires imaging data of a subject in an examination region. A reconstruction

30   processor reconstructs the acquired imaging data into an image representation. A pair of electrodes are adapted to contact a thoracic region of the subject. An electrical meter measures a time-varying electrical parameter across the electrode pair. A monitor extracts a time-varying respiration characteristic from the measured time-varying electrical parameter.

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According to another aspect, a medical diagnostic imaging method is provided. Imaging data of a medical imaging patient is acquired. At least a part of the acquired imaging data is reconstructed into an image representation. A time-varying electrical parameter is measured across an electrodes pair during the acquiring of imaging data. A time-varying respiration characteristic is computed based on the measured time-varying electrical parameter.

One advantage resides in providing respiratory gated medical diagnostic imaging without encumbering the imaging patient with a bulky and uncomfortable electromechanical transducer.

Another advantage resides in providing an electronic signal indicative of aspects of respiratory activity such as inhalation or a breath hold.

Yet another advantage resides in providing respiration information during cardiac computed tomography without contacting the patient with additional probes.

Numerous additional advantages and benefits will become apparent to those of ordinary skill in the art upon reading the following detailed description of the preferred embodiments.

The invention may take form in various components and arrangements of components, and in various process operations and arrangements of process operations. The drawings are only for the purpose of illustrating preferred embodiments and are not to be construed as limiting the invention.

FIGURE 1 diagrammatically shows a computed tomography imaging system including *in situ* respiratory cycle monitoring.

FIGURE 2 diagrammatically shows a suitable embodiment of the impedance meter of the computed tomography imaging system of FIGURE 1.

FIGURE 3 diagrammatically shows a suitable embodiment of the respiratory monitor of the computed tomography imaging system of FIGURE 1.

FIGURE 4 diagrammatically shows another suitable embodiment of the respiratory monitor of the computed tomography imaging system of FIGURE 1.

With reference to FIGURE 1, a computed tomography imaging system 10 includes a computed tomography scanner 12. Although the invention is described with exemplary reference to computed tomography, it will be appreciated that the invention will also find application in conjunction with other medical diagnostic imaging systems that employ other imaging modalities

such as single photon emission computed tomography, positron emission tomography, magnetic resonance imaging, ultrasound imaging, and the like.

The computed tomography scanner 12 includes an x-ray source 14 and a receiving x-ray detector array 16 oppositely arranged respective to an imaging region 20 defined by a rotating gantry 18. A patient to be imaged is arranged on a subject support 22 and inserted into the imaging region 20. The x-ray source 14 produces a cone-shaped, wedge-shaped, or otherwise-shaped x-ray beam directed into the imaging region 20, where x-rays are partially absorbed by the patient. The absorption-attenuated x-ray intensities are measured at the x-ray detector array 16 after traversing the imaging region 20, and x-ray absorption data generated by the detector is stored in an imaging data memory 24.

Preferably, the gantry 18 rotates continuously during the imaging to acquire projection views over at least a 180° angular span. In helical computed tomography imaging, the subject support 22 advances linearly during the imaging. The combination of linear movement of the subject support 22 and circular rotation of the gantry 18 produces a helical orbiting of the x-ray source 14 about the imaging region 20, during which three-dimensional volumetric imaging data is acquired and stored in the imaging data memory 24.

Prior to the imaging, a pair of electrodes 30, 32 are electrically and mechanically connected to a thoracic region of the patient. Preferably, the electrodes 30, 32 are arranged with a substantial portion of the thoracic region therebetween. The electrodes 30, 32 are suitably made of a material or materials with low x-ray absorption characteristics. Similarly, for other imaging modalities the electrodes are preferably constructed to minimize interference with imaging. For magnetic resonance imaging applications, for example, the electrodes should be made of a non-magnetic material or materials.

The electrodes 30, 32 electrically communicate with an impedance meter 34 that measures an impedance across the electrodes 30, 32. The measured impedance can include resistance, capacitance, inductance, reactance, or a complex impedance combining resistance and reactance. The measured impedance varies with spatial separation of the electrodes 30, 32. This spatial separation changes as the thorax expands and contracts with inhalation and exhalation, and so the measured impedance carries respiratory cycling information. An impedance function is measured, which varies with time as the respiratory cycling progresses.

A respiration monitor 36 receives the measured impedance function and computes one or more respiratory characteristics. For example, a periodicity of the impedance function tracks the respiration rate. A slope of the impedance function relates to respiration state. If the impedance is a resistance, the resistance increases with increasing tidal volume of air in the lungs because the path length between the electrodes 30, 32 increases. Hence, inhalation is indicated by a positive

slope of the time-varying resistance function. Exhalation is similarly indicated by a negative slope of the resistance function. A breath-hold or other extended interval between inhalation and exhalation is indicated by a generally horizontal slope. Extrema (i.e., peaks and valleys, or maxima and minima) of the time-varying impedance function indicate transitions between inhalation and exhalation, and provide temporal markers for estimating respiratory cycle phase.

The one or more respiratory characteristics output by the respiration monitor **36** can be used in various ways. For example, in one application a respiratory cycle binning processor **40** sorts acquired projection views by respiratory cycle phase (ϕ) which is indicated by the respiratory monitor **36**. The respiratory cycle binning processor **40** sorts the projection views into respiratory cycle phase bins **42**, with each bin storing projection views acquired over a selected respiratory cycle phase interval. To meet data sampling requirements, some data may be from another phase which is similarly configured. A respiratory cycle phase selector **44** selects one or more respiratory phases for reconstruction. A reconstruction processor **46** reconstructs imaging data stored in the selected respiratory cycle phase bins **42** into one or more volumetric image representations that are stored in an image memory **48**. By reconstructing data acquired over a limited respiratory cycle phase interval, image motion blurring due to respiration is reduced. A video processor **50** generates a displayable image from the volumetric image representation, such as a maximum intensity projection, an extracted two-dimensional slice, or a three-dimensional rendering, which is displayed on a computer **52** or other user interfacing device.

In another application, the respiratory characteristic is used to identify respiratory activity during imaging data acquisition. An image registration processor **60** relatively spatially registers image representations based on the monitored respiratory activity to correct for respiration-related motion of an imaged organ of interest. For example, portions of the reconstructed image are expanded or contracted to adjust the images to a constant respiratory state. The video processor **50** produces displayable images of the spatially registered image representations, which have reduced respiration-related motion blurring due to the image registration processing.

In yet another application, the respiratory characteristic is used to trigger image data acquisition or to perform prospective respiratory gating of the data acquisition. For example, the respiratory characteristic can be monitored to detect a breath-hold state. When a breath-hold is detected, an imaging sequence supplied to a computed tomography (CT) scanner controller **64** via the user interface **52** is triggered. In response to the triggering, the controller **64** causes the computed tomography scanner **12** to execute the selected imaging sequence. In prospective respiratory gating, the controller **64** initiates intervals of imaging data acquisition during selected

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respiratory cycle phase intervals indicated by the respiration monitor 36. Dose-modulated imaging synchronized with the respiratory cycle can similarly be performed.

For cardiac computed tomography imaging, an electrocardiograph 66 suitably monitors cardiac cycling simultaneously with monitoring of the respiratory cycle using the electrodes pair 30, 32. In a preferred embodiment, the impedance meter 34 measures a pulse-modulated signal with a pulse frequency substantially higher than the heart rate. Hence, the impedance measurement signal and the electrocardiographic signal are readily decoupled by frequency selective filtering. The electrocardiograph 66 in the illustrated embodiment gates image data acquisition. Alternatively, the electrocardiographic information is conveyed to the binning processor 40 to bin the data based on both cardiac and pulmonary phase.

With reference to FIGURE 2, the exemplary impedance meter 34 measures a time-varying resistance R(t) 70. Specifically, a pulse generator 72 produces a voltage pulse train applied across the electrodes pair 30, 32. The pulse train is preferably has a frequency in the tens of kilohertz range, with the precise frequency selected to lie within applicable frequency allocation bands to avoid generating undesirable radio frequency interference. The frequency of the pulse generator 72 is optionally adjustable, in which case the applied frequency can be optimized to maximize the impedance signal. An amplitude of the voltage pulses is typically a few volts or lower. An ammeter 74 measures current at the pulse train frequency. Optionally, a frequency-selective filter 76 removes noise or other interference from the measured current. An impedance calculator 78 computes the resistance by ratioing applied voltage and measured current to compute the time-varying resistance R(t) 70.

Those skilled in the art can readily modify the exemplary impedance meter 34 to suit specific applications or to take advantage of available electronic components. For example, a current pulse train can be sourced, and voltage measured by voltmeter. Other types of impedance such as capacitance, inductance, or complex impedance can be computed by suitably taking into account phase-shifts between the voltage and current pulses or by otherwise combining the voltage and current data. Moreover, a voltage, current, or other electrical parameter can be used to characterize respiration. For a steady-state voltage pulse train, the output of the ammeter 74 carries respiratory information. For respiratory monitoring during magnetic resonance imaging, a low frequency or d.c. voltage or current is preferably applied to obviate radio frequency interference concerns.

When using the voltage pulse train input produced by the pulse generator 72, it will be appreciated that the electrocardiograph 66 can measure an electrocardiographic signal 80 simultaneously with measurement of the time-varying resistance 70 or other time-varying electrical parameter. Using the electrode pair 30, 32 for both respiratory monitoring and cardiac monitoring

advantageously reduces imaging delays and simplifies patient preparation, and further benefits the patient by reducing a total number of contacting probes. To remove the high-frequency signal components produced by the pulse generator 72, a lowpass filter 82 is suitably applied prior to electrocardiographic measurement. Although a separate filtering component 82 is shown in

5   FIGURE 2, the filter is optionally omitted if a frequency response of the electrocardiograph 66 is such that the electrocardiograph 66 does not respond to signal components at the pulse train frequency.

With reference to FIGURE 3, an exemplary embodiment of the respiration monitor 36 outputs a respiration state 90 which classifies the patient as being in one of an exhalation state,

10   an inhalation state, and a breath-hold or transitional respiratory state. A first derivative processor 94 computes a first derivative of the time-varying resistance 70 using an analog differentiator circuit, numerical differentiation, analytic differentiation of a fitted curve or fitted spline segments, or the like. A threshold processor 96 classifies the first derivative signal as one of positive, negative, or substantially zero. It will be recognized that a positive first derivative of the measured resistance 70

15   corresponds to an increasing time-varying resistance, which in turn corresponds to an increasing resistance path length that occurs during inhalation chest expansion. Similarly, a negative first derivative of the measured resistance 70 corresponds to a decreasing time-varying resistance, a decreasing path length, and hence exhalation. A substantially zero first derivative of the measured resistance 70 corresponds to a constant thoracic volume, which typically indicates an intentional

20   breath-hold by the patient or a fully inhaled or fully exhaled state.

With continuing reference to FIGURE 3, the respiration monitor 36 also preferably measures a respiration rate 98. The respiration rate is relatively low frequency, typically around 12 breaths per minute (~0.2 Hz). In one suitable measurement approach a low pass filter 100, such as a filter with a cutoff frequency f_c~0.5 Hz, filters out higher frequency components of the

25   time-varying resistance R(t) 70. A frequency processor 102 determines a temporal frequency of the filtered signal corresponding to the respiration rate 98 using a fast Fourier transform (FFT) analysis, peak detection and counting, or the like.

Other respiratory parameters can similarly be estimated from the time-varying resistance R(t) 70. For example, a respiratory cycle phase processor 104 estimates a respiratory

30   cycle phase $\phi(t)$ 110 based on the time-varying resistance R(t) 70 and/or time-varying functions extracted therefrom such as the first derivative generated by the first derivative processor 94 or a second derivative generated by a second derivative processor 106. To reduce computational complexity, it is typically advantageous to differentiate the output of the first derivative processor 94 to obtain the second derivative, as shown in FIGURE 3. However, a method that computes the

35   second derivative directly from the time-varying resistance 70 can also be employed. The

respiratory cycle phase $\phi(t)$ **110** is suitably estimated by the respiratory cycle phase processor **104** by curve-fitting the time-varying resistance R(t) **70** and/or derivatives thereof with a suitable respiratory cycle model. Alternatively, the respiratory cycle phase processor **104** can locate critical points (such as maxima, minima, or inflection points) of the time-varying resistance R(t) **70** and/or

5   derivatives thereof that correspond to certain respiratory cycle phases such as transitions from inhalation to exhalation, transitions from exhalation to inhalation, and the like.

With reference to FIGURE 4, another exemplary embodiment of the respiration monitor **36** outputs quantitative values for tidal volume V(t) **120** of air in the lungs. Specifically, a patient calibration **122** that correlates tidal volume with resistance between the electrodes **30, 32** is

10   obtained prior to tomographic imaging, for example by measuring resistance values across the electrodes **30, 32** during quantitative spirometric measurement of tidal volumes. Using the patient calibration **122**, the time-varying resistance R(t) **70** is processed by a tidal volume transform processor **124** to convert the resistance R(t) **70** to quantitative values of tidal volume V(t) **120**.

It will be appreciated that components of the respiration monitor of FIGURES 3

15   and 5 are readily combinable. Moreover, other respiration characteristics besides the exemplary respiration state, respiration rate, respiration cycle phase, and tidal volume characteristics can be computed based on the time-varying resistance or other time-varying electrical parameter **70**. Again, the respiration rate can be fit to a model and used to predict future respiratory cycles.

The invention has been described with reference to the preferred embodiments.

20   Obviously, modifications and alterations will occur to others upon reading and understanding the preceding detailed description. It is intended that the invention be construed as including all such modifications and alterations insofar as they come within the scope of the appended claims or the equivalents thereof.